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Developing a Three Dimensional Finite Element Model of the Anterior Cruciate Ligament to Examine the Risk Factors for Women during the Sidestep Cutting Maneuver

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DEVELOPING A THREE DIMENSIONAL FINITE ELEMENT MODEL OF THE ANTERIOR CRUIATE LIGAMENT TO EXAMINE THE RISK FACTORS FOR WOMEN DURING THE SIDESTEP CUTTING MANEUVER

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Abstract

Anterior cruciate ligament injuries (ACL) have been increasing throughout the years and close to one billion dollars are spent annually on ACL injuries. What is also alarming is that women are 4 to 6 times more susceptible to ACL injuries than men. Many researchers have proposed risk factors that are related to the higher incident rates seen in women. However, there currently isn't enough research to verify these risk factors. It is important to find prevention methods to reduce the instances of ACL injuries. The goal of this project is to examine how proposed risk factors affect the ACL during a sidestep cutting maneuver. A three dimensional finite element model of the ACL and knee was developed and combinations of loads are applied. Using the finite element analysis program ANSYS, the stresses on the modeled ACL are examined. The model is tested against risk factors that are proposed to be related to the increase in incident rates in women compared to men, such as difference in ligament size and quadriceps angle.

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1. Introduction

1.1 Objective

The objective of this project is to develop a three dimensional finite element model of the anterior cruciate ligament to study the effect of varying the range of values of intrinsic risk factors on the stress acting on the ACL during the sidestep cutting maneuver. This project examines the correlation between risk factors and ACL incident rates in women.

1.2 Rationale

Anterior cruciate ligament (ACL) injuries are among the most common ligament injuries and the most common major knee injury suffered by athletes (Gäbler 2004). With an estimate of 80,000 ACL tears occurring annually in the United States, an equivalent of almost one billion dollars are being spent on ACL injuries annually (Griffin et al., 2000). This amount will continue to increase, as shown in the last 15 years, knee ligament injuries have increased by 172% (Gäbler 2004). With the incidences of ACL injuries increasing with time it is important to find prevention methods to reduce the number of instances as well as the money being spent on ACL injuries. By developing a three dimensional finite element model can help identify future ACL injury prevention methods and reduce the incident rates and money being spent.

ACL injuries often occur during sports for both contact and non-contact situations, where seventy percent of ACL injuries occur in non-contact situations (Griffin et al. 2000). These non-contact situations include landing or one of the stance phase of “high risk” sporting postures such as sidestepping (Besier et al. 2001), both which are key offensive strategies seen in sports such as basketball, handball, and soccer (McLean 2004). The three dimensional finite element model of this maneuver can help future studies in understanding the relationship between varying intrinsic risk factors and sidestepping.

Another alarming fact is that women are at a higher risk of ACL injuries. Women are 4 to 6 times more susceptible to ACL injuries than men (Hewett et al. 2005). However the reasons behind this have yet been fully understood. By testing different risk factors using a finite element model, the model may help reinforce the understanding of the correlation between different intrinsic factors and ACL risk rates in women.

1.3 State of the Art

1.3.1 Risk Factors of ACL Injuries

Studies have been done to identify patterns in ACL injuries and a variety of intrinsic and extrinsic factors were proposed as contributing to the increase in ACL injury rate in women. The intrinsic factors include ACL size, intercondylar notch size, quadriceps angle, joint laxity, and hormonal variations (Chappell et al. 2002). The extrinsic factors include body movement, shoe-surface interface, skill level (Arendt et al. 1999), muscle strength, level of conditioning, and motor control strategies (Chappell et al. 2002). A table of these risk factors is organized in Table 1.

Such patterns have also been studied between men and women. Female athletes have smaller knee flexion angles, greater knee valgus angles, increased quadriceps muscle activation, and decreased hamstring muscle activation during cutting tasks compared to male athletes (Malinzak et al. 2001). These proposed differences may be the cause of higher incident rates in women because they tend to increase the strain on the ACL (Chappell et al. 2002).

Table 1: Intrinsic and Extrinsic Risk Factors (Arendt et al. 1999 & Chappell et al. 2002)

Intrinsic Risk Factors	Extrinsic Risk Factors
Ligament Size	Level of Muscle Strength
Intercondylar Notch Size	Level of Conditioning
Quadriceps Angle ("Q" Angle)	Motor Control Strategies
Knee Laxity	Shoe-Surface Interface
Hormonal Variations	Skill Level

1.3.2 Sidestep Cutting Maneuver

Studies have been done to examine the sidestep cutting maneuver (Figure 1) as it is a common offensive strategy in athletic plays (McLean et al. 2004). A study that gives insight on the biomechanics of sidestep cutting is by McLean et al. (2004) who studied the effect of gender and defensive opponent on the biomechanics of sidestep cutting. The 3D motion and ground reaction forces of eight male and eight female subjects who performed sidestep cutting maneuvers were recorded. Their results show that females had less hip and knee flexion, hip and knee internal rotation, and hip abduction. Females had higher knee valgus and foot pronation angles, and increased variability in knee valgus and internal rotation. Increased medial ground reaction forces and flexion and abduction in the hip and knee occurred with the defensive player for both genders. These differences may suggest that the positioning of the body is connected to ACL incident rate. In a different study also done by McLean et al. (2004), they studied the knee joint sagittal plane forces on the ACL during sidestep cutting in 10 male and 10 female subjects. The result from their study suggests that the valgus loads on the ACL were high enough to rupture the ligament and it occurs more frequently in women.

Figure 1: Demonstration of the Sidestep Cutting Maneuver

1.3.3 Force Combination of Sidestep Cutting Maneuver in the Knee

The sidestep cutting maneuver involves a combination of different forces. The forces during this maneuver that affect ACL loading include the quadriceps, gastrocnemius, and hamstring muscle forces (McLean et al. 2003), flexion/extension moment, varus/valgus moment, and internal/external rotation moment (Besier et al. 2001).

The moments about the knee during the sidestep cutting maneuver involves the three rotational degrees of freedom, the flexion/extension, varus/valgus, and internal/external rotation (Figure 2). In Besier et al.'s (2001) study on the external loading of the knee joint during cutting maneuvers, their study recorded the moments about these three rotational degrees of freedom.

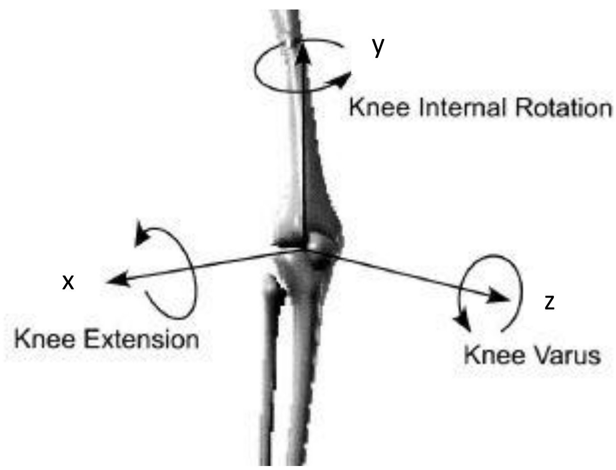

Figure 2: Kinematic Model of Three Rotational Degrees of Freedom of the Right Knee

The results from their study show that there are three stages of stance phase. Besier et al. (2001) classified them as the weight acceptance, the peak push off, and the final push off phase. Four different sets were tested, the sidestep cutting maneuver at a 30 and 60 degree angle, a straight run, and a crossover cutting maneuver at 30 degrees. Besier et al.'s study shows that flexion, varus, and internal rotational moments are involved in a sidestep cutting maneuver.

To further understand how the three moments impact the forces in the ACL, Markolf et al.'s (1990) study on the direct measurement of resultant forces in the ACL shows the resultant forces in the ACL during passive flexion/extension, varus/valgus moment, and internal/external torque, individually. The results from their study show that that as the knee flexion angle decreases, the force on the ACL increases. The varus moment at lower flexion angles yield greater forces on the ACL compared to valgus moment. The varus moment at different flexion angles also generate varying forces compared to the valgus moment. The results also show that the internal torque at lower flexion angles yield greater forces on the ACL than applying external torque.

1.3.4 Finite Element Models

With the available references to the mechanical properties and dimensions of the ACL, researchers are able to develop various types of models to study the ACL without having the difficulty of obtaining it experimentally using actual human ACL. Currently, three dimensional finite element models have been used to study the ACL and only a few full three dimensional finite element models have been developed (Limbert et al. 2004). According to Park et al. (2010), the finite element (FE) method is an effective approach in identifying stress distributions in the ACL in reaction to loading and tibiofemoral movements.

The use of three dimensional finite element models (FEM) to study the ACL have been demonstrated by researchers, a list of studies of some of those who have used finite element models can be found in Appendix A.

1.4 The Approach

In order to study the impact of different risk factors on the ACL during a sidestep cutting maneuver, a 3D finite element model of the ACL is made. The construction and assembly of the model is done in SolidWorks and the FEA is done using the ANSYS software. And the necessary parameters for the model are obtained from other studies. Similar to Song et al.'s study in 2004, the 3D finite element model analyzes the stress on the ACL. The risk factors are examined on the model similar to Park et al.'s (2010) study where the intercondylar notch was examined using the FEM of the ACL. Although there are many proposed risk factors this model aims to study two of these which are ligament size and quadriceps angle. A list of risk factors and studies done relating to the corresponding risk factors can be found in Appendix B.

In the current state of the art, there isn't a finite element model of the knee that examines the sidestep cutting maneuver. This model incorporates this maneuver because it is the most common injury suffered by athletes in non-contact situations. To simulate the sidestep cutting maneuver, a combination of forces is applied onto the model using the results from the study done by Besier et al. (2001).

The finished model is then evaluated by comparing the results of an anterior tibial load test from Peña et al.'s (2006) study and the stresses on the ACL is examined.

2. Method

2.1 Generation of Finite Element Model of the ACL

The finite element model includes the femur, tibia, the ACL and the PCL. The 3D geometry of the femur and tibia was obtained from the CT data of a frozen cadaver made available by the Pacific Research Labs. The 3D geometry of the ACL and PCL was modeled using ProENGINEER Wildfire. To construct a three dimensional model of the two ligaments, assumptions were made and the mechanical properties, anatomical properties and dimensions were applied.

2.1.1 Assumptions

For this model two assumptions are made and they are as follows,

1. The ligament and bone material are considered isotropic. Therefore the material properties of the body and insertion site of the ACL are the same.
2. The viscoelasticity, creep, and relaxation are neglected due to high ratio between viscoelastic time constant and loading time.

2.1.2 Mechanical Properties

There are two main types of materials involved in this FEM. They are the bone and ligament. The mechanical properties for both these materials were applied. The femur and tibia were applied with the bone material. The ACL and PCL were applied with the ligament material. The chart with the bone and ligament mechanical properties is shown in Table 2.

Table 2: Mechanical Properties of the Bones and Ligament (Ozkaya & Nordin 1999)

Properties	Materials	
	Bone	Ligament
Density (kg/m^3)	1900	---
Yield Strength (MPa)	80	---
Ultimate Strength (MPa)	130	70
Elastic Modulus (GPa)	17	0.40
Shear Modulus (GPa)	3.3	---
Poisson's Ratio	0.40	0.40

In Chandrashekar et al.'s (2006) study in the differences in tensile properties of the human ACL, they were able to test multiple human ACLs from both male and female cadavers, their results were compared with the ACL studies of Woo et al. (1991) and Noyes and Grood (1976). From Chandrashekar et al.'s study, noticeable differences in male and female ACLs have been seen. In mechanical properties, the modulus of elasticity and stiffness is much greater in male than female ACL. The stress, strain, and load at failure are seen to be greater in male than female ACL. For the dimensions of the ACL, the length and cross sectional area of the male ACL is also greater than the female ACL.

The additional mechanical properties of the model ACL are based on the mechanical properties measured by Chandrashekar et al (2006), shown in Table 3. And the mechanical properties of the PCL are obtained from Prietto et al. (1988) shown in Table 4.

Table 3: Mechanical Properties of the ACL

Study	Elongation at failure (mm)	Strain at failure	Load at failure (N)	Stress at failure (MPa)	Stiffness (N/mm)	Modulus of elasticity (MPa)
Chandrashekar et al. 2006	8.95 ± 2.12	0.30 ± 0.06	1818 ± 699	26.35 ± 10.08	308 ± 89	128 ± 35

Table 4: Mechanical Properties of the PCL

Study	Strain at failure (%)	Load at failure (N)	Stress at failure (MPa)	Stiffness (N/mm)	Modulus of elasticity (MPa)
Prietto et al. (1988)	28.5	1627	26.8	204	109

2.1.3 Anatomical Properties

In order to create an accurate knee model the anatomical properties need to be applied. To analyze the ACL the location in which the ACL and PCL attaches to the femur and the tibia is obtained and shown in Table 5. In Zantop et al.'s (2006) study on ACL anatomical reconstruction; they reported the anatomical properties of the ACL such as the femoral and tibial insertion and origin of the ACL. The anatomical properties applied to the model for the ACL are based on Zantop et al.'s (2006) study. And the PCL is based on Mejia et al. (2002), Girgis et al. (1975), and Sheps et al. (2005).

Table 5: Anatomical Properties of the ACL

Ligament	Study	Femoral Origin & Insertion	Tibial Insertion
ACL	Zantop et al. 2006	- The femoral origin of the ACL is at the lateral femoral condyle. - Using the quadrant method, the center of the ACL attachment on the lateral femur condyle can be located at 24.8% of the distance defined by the intersection of Blumensaat's line and at 28.5% of the height of the lateral femoral condyle from Blumensaat's line.	- Tibial insertion is approximately 120% of the area of the femoral insertion site. - The insertion site for the ACL on the tibia is located in the area between the medial and lateral tibial spine.
PCL	Mejia et al. 2002	- Femoral insertion site is defined using the clock marking technique.	--
	Girgis et al. 1975	- The femoral origin is a broad crescent-shaped area on the anterolateral aspect of the femoral condyle.	--
	Sheps et al. 2005	--	- Tibial insertion is similar to a shape of a trapezoid - The insertion is located in a fossa between the posterior aspects of the medial and lateral plateau of the tibia

2.1.4 Dimensions

The ACL was modeled using the dimensions measured by Chandrashekar et al.'s (2005) study. A table of the measurements is shown in Table 6. And the PCL was modeled using the dimensions measured by Harner et al. (1999) and Prietto et al. (1988) shown in Table 7.

Table 6: Dimensions of the ACL

Study	Length (mm)	Mid-substance Area (mm^2)	Femoral Insertion Width (mm)	Femoral Insertion Length (mm)	Tibial Insertion Width (mm)	Tibial Insertion Length (mm)
Chandrashekar et al. 2005	29.82 ± 2.51	83.54 ± 24.89	--	--	--	--
Zantop et al. 2006	--	--	11	18	11	17

Table 7: Dimensions of the PCL

Study	Length (mm)	Mid-substance Area (mm^2)	Femoral Insertion Area (mm^2)	Tibial Insertion Area (mm^2)
Harner et al. (1999)	--	50	128	153
Prietto et al. (1988)	31.4	--	--	--

2.1.5 Sidestep Cutting Maneuver Parameters

Once the mechanical and anatomical properties are applied to the model, forces are applied to simulate the sidestep cutting maneuver. These forces include the loads from muscle activation reported in McLean et al.'s (2003) study. Although the sidestep cutting maneuver involves various muscle groups the three major groups that affect the ACL and are applied onto the model are the quadriceps (Rectus femoris, vastus lateralis, vastus medialis and vastus interm), gastrocnemius, and hamstrings (Li et al. 1999 & Dürselen 1995). The moments that occur in the knee during the sidestep cutting maneuver observed by Besier et al.'s (2001) are also applied. The moments at peak push off phase during sidestep cutting are as follows, a flexion moment of 174.3 Nm, and varus and internal rotation moment of 24.9 Nm.

2.1.6 Boundary Conditions

Boundary conditions are then applied onto the model for finite element analysis. The femur is rigidly fixed as the tibia is free to move in the flexion plane, as well as the varus/valgus and internal/external rotations. Three contact zones are also applied. First is a frictionless contact zone set between the base of the femur with the cranial portion of the tibia. Second is a bonded contact zone between the ACL and the femoral insertion site. The third is also a bonded contact zone between the ACL and the tibial insertion site.

2.2 Major Challenges

Through the process of this project couple of obstacles was encountered. There were issues concerning the meshing process of the model. The first constructed CAD model of the knee using the bone models made by Viceconti (2009) using the CT scan of frozen cadavers from National Library of Medicine received errors in ANSYS refusing to mesh the model. Considerable amount of time was used in attempt to solve this issue because the analyses cannot proceed without the model successfully meshed. Explanations and attempts to solve this issue can be found in Appendix C. This issue has not been officially solved however a second set of bone models was acquired from the Pacific Research Labs (Papini 2009) and assembled. The second model was able to mesh successfully without any issues. This lead to the possible conclusion that the problem originated from the first CAD model and ANSYS was not able to mesh that particular CAD assembly.

Another major issue that occurred during this project is the application of muscle loads onto the model. Initially muscle activation loads that simulated the sidestep cutting maneuver was one of the risk factors that were to be tested. However, due to time constraints the decision was made to not test the muscle loads but instead test the other two risk factors (ligament size and quadriceps angle). The problem that occurred happened after applying the suggested muscle activation loads by McLean et al. (2003) found in Appendix D. Parts of the knee model became distorted and the stress values obtained were abnormally high. Further troubleshooting was necessary to figure out why this happened. Possible causes to this problem may be due to inappropriate mechanical properties or magnitude of muscle loads.

2.3 Finite Element Analyses

With the mechanical and anatomical properties, forces, and boundary conditions applied onto the model, finite element analyses were performed. Each analysis tests a different risk factor.

These analyses were conducted using ANSYS Workbench v.13 software. And the steps in applying the risk factors are as follows.

2.3.1 Applying Risk Factors

To identify the correlation between certain risk factors and the incident rates of female athletes, risk factors such as ligament size, quadriceps angle and muscle strength was examined. Each risk factor was applied to the model and tested against different knee flexion angles (0°, 30°, 60°, 90°, and 120°).

2.3.1.1 Ligament Size

To examine the stress on the ACL according to size, the size of the ACL was varied in increments of 3mm. According to Chandrashekar et al. (2005), the average size of the male ACL is 29.82mm and 26.85mm for females. The sizes of ligament tested were in the range of 24-33mm. This range is chosen to test ligament sizes greater and less than the average ligament sizes of male and female ACLs.

2.3.1.2 Quadriceps Angle

The normal quadriceps angle in men is 14 degrees and 17 degrees in women (Conley et al. 2007). To determine if an increase in quadriceps angle has an impact on the ACL, the quadriceps angle was tested in the range of 15-30°, in increments of 5°. To test different quadriceps angles, the fixed femur was repositioned according to the quadriceps angle before testing.

3. Results & Discussion

After the appropriate settings such as the mechanical properties and boundary conditions are applied and set up, the finite element model of the knee was meshed (Figure 3). A total of 40 different setups were constructed and the varied ligament length and Q-angles at different knee flexion angles were tested. The results were then recorded. The model was also evaluated against another model by applying an anterior tibial load and compressive load at full extension of the knee.

3.1 Finite Element Model of the Knee

The assembled CAD model of the knee inputted into ANSYS was setup and meshed to conduct finite element analyses. Notice that the generated number of elements for the femur is less than the tibia for the meshed model. This is because the original CAD model of the femur consists of fewer faces than the tibia. Also the number of elements is reduced due to the extensive run time it requires to generate the mesh and run analyses. The greater the number of elements the model consists of the longer the run time.

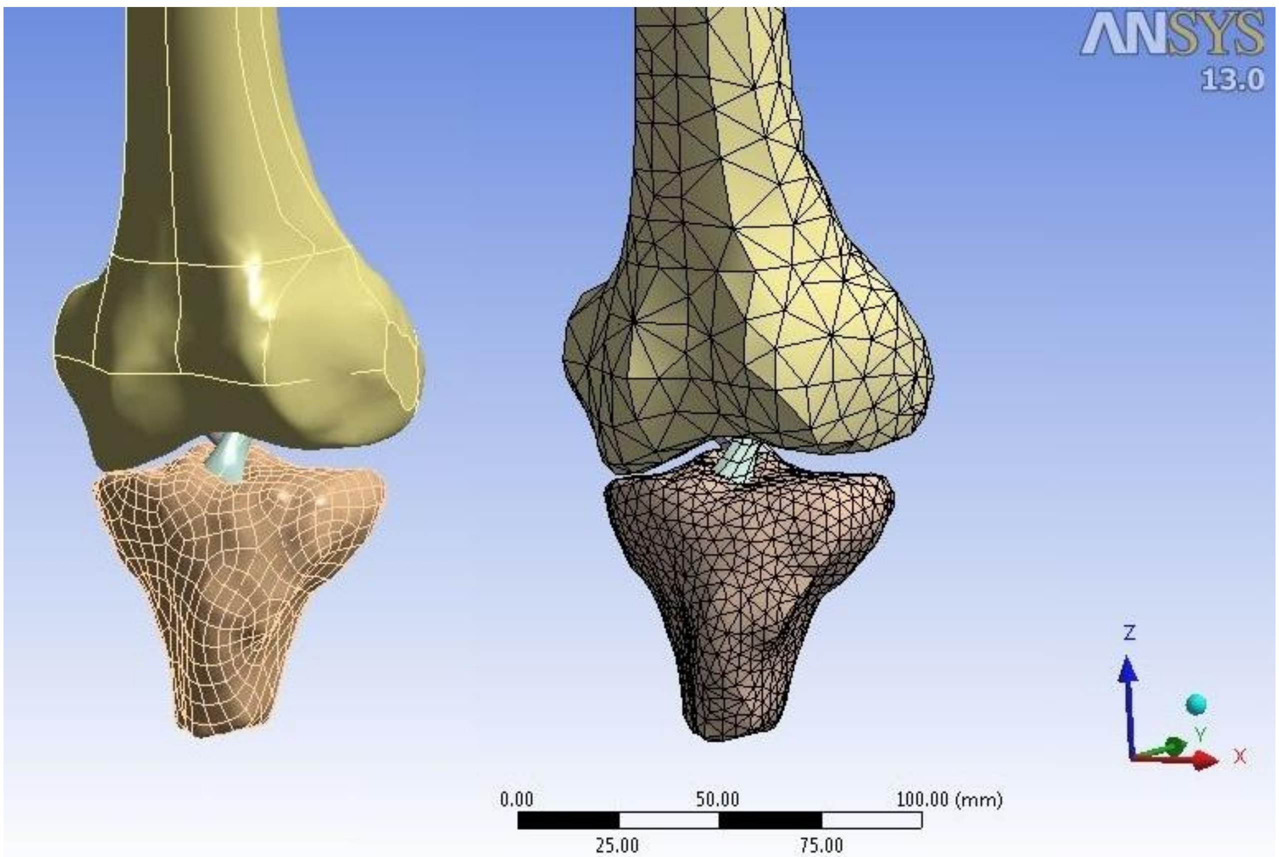

Figure 3: The Assembled Knee Model (Left) and the Generated Meshed Model (Right)

3.2 Evaluation of the Finite Element Model

To evaluate the model, the model was compared with the model developed by Peña et al. (2006). The study applied a 134N anterior tibial load and a compressive load of 1150N at full extension of the knee. Identical loads were applied onto the constructed model. The results were then compared with the results from Peña et al.'s study (2006). The results did not turn out to be the same. The results obtained for the maximum principal stress on the ACL were lower than the values obtained from Peña et al.'s study. Peña et al.'s study reported an average maximum principal stress of around 6.5 MPa and a maximum of 15 MPa. The results from my model showed an average of 2.5 MPa and a maximum of 4.7 MPa (Figure 4).

The difference in result values seen in the comparison with Peña et al.'s study (2006) may be due to the lack of anatomical structures. Peña et al.'s model incorporates all four knee ligaments as well as the meniscus. This may cause loads on the ACL to be shared by other ligaments. The mechanical properties used for Peña et al.'s study may have been different from the ones obtained for this model. Another reason for the difference in value may be because of the ACL model used. The ACL model used in Peña et al.'s study was generated using data collected from magnetic resonance imaging (MRI) while the ACL model in this study was modeled in ProENGINEER using parameters and information proposed by Chandrashekar et al. (2005) and Harner et al. (1999).

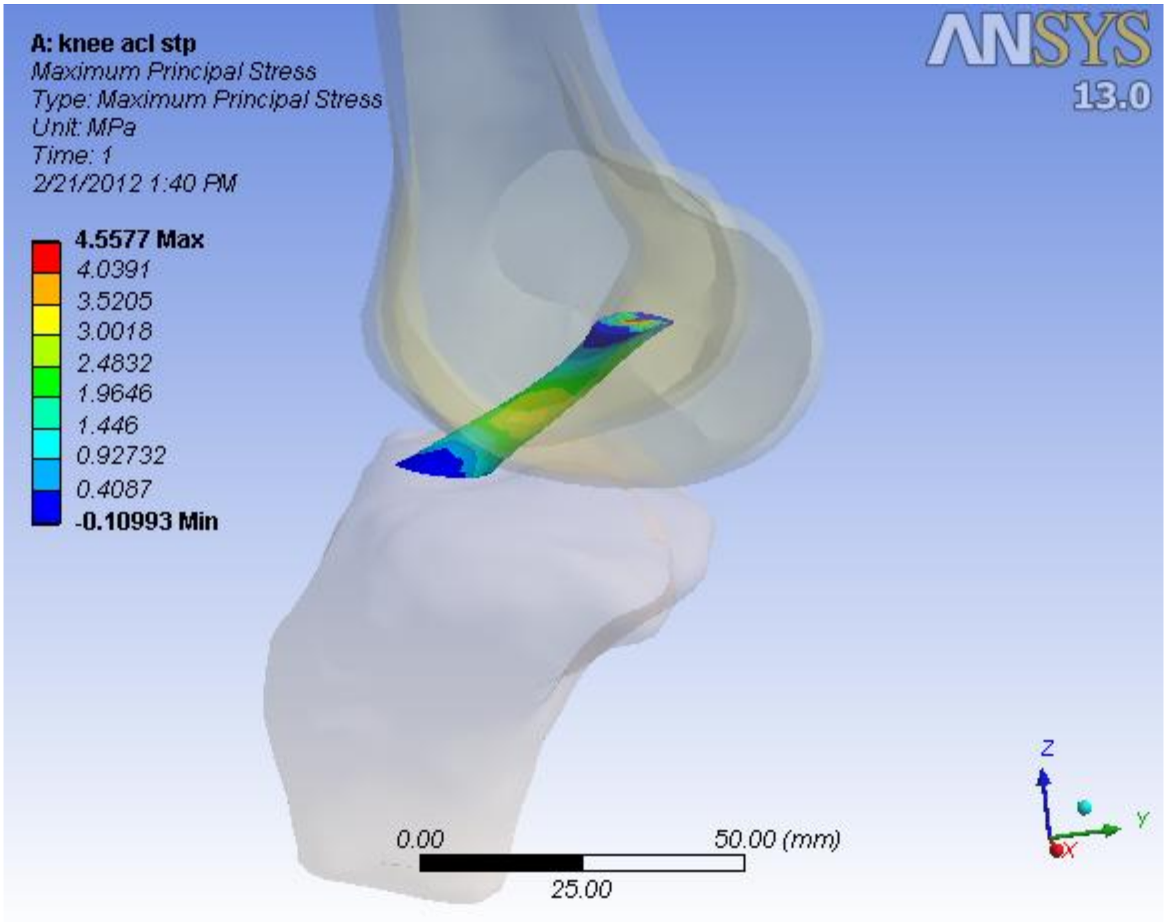

Figure 4: The Finite Element Model Subjected to 134N of Anterior Tibial Load and 1150N of Compressive Load

3.3 Stress Analysis

Finite element analysis was done on the model and the two risk factors (ligament size and quadriceps angle) were varied along with the knee flexion angle (0° to 120°). Ligament length ranging from 24 to 33 mm and quadriceps angle ranging from 15° to 30° were analyzed.

3.3.1 Ligament Size

The results from analyzing various ligament lengths are shown in Table 8. The maximum principal stresses were recorded and plotted in Figure 5.

Table 8: The Recorded Maximum Principal Stresses from Analyzing Various Ligament Sizes

Ligament Size (mm)	Max Principal Stress (MPa) Knee Flexion Angle (degrees) 0	30	60	90	120
24	2.66490	1.54840	0.00037	0.25986	0.00044
27	3.37690	3.06120	0.00056	0.06041	0.00102
30	3.12510	2.08080	0.00361	0.61735	0.00092
33	2.05060	1.58740	0.00362	0.05600	0.00054

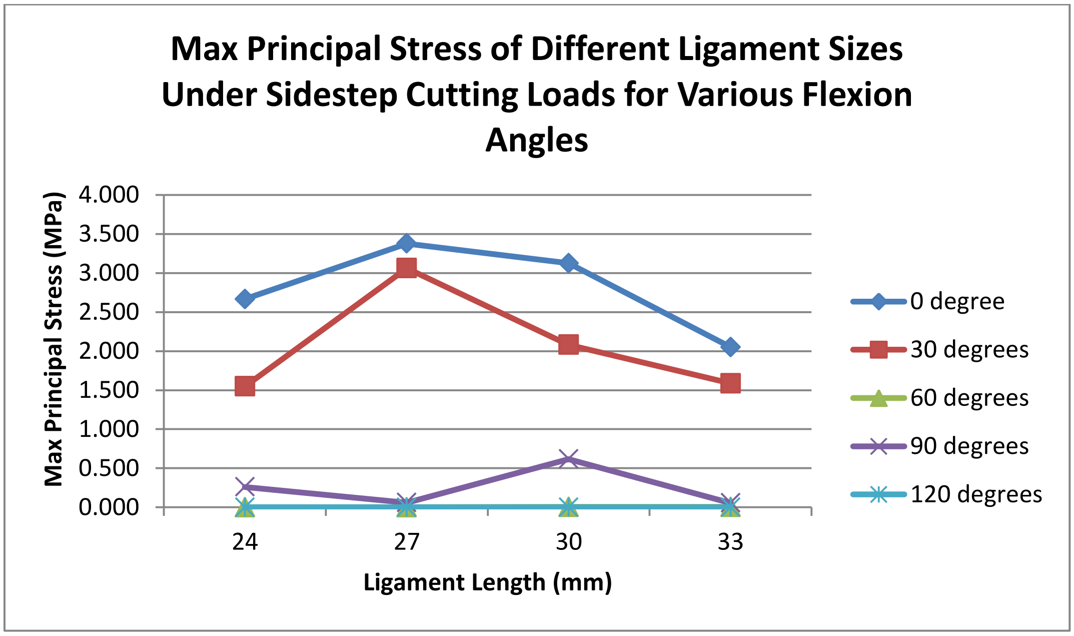

Figure 5: Graph of the Maximum Principal Stresses Recorded from Analyzing Various Ligament Sizes at Different Flexion Angles

Looking at Figure 5, it can be seen that the greatest maximum principal stress is experienced at a knee flexion angle of 0°. And the second greatest at 30° while the stress at other knee flexion angles (60°, 90° & 120°) are significantly less. The results from this test agrees with what Markolf et al.'s study (1990) which suggests that as knee flexion angle decreases, the stress on the ACL increases. Possible reason for the ACL to experience less stress at a greater flexion angle may be due to the relaxation of the ligament as the femoral condyles are flexed. The femoral insertion site of the ACL is not located at the center of rotation of the knee. Therefore the displacement of the femoral insertion site as the knee is flexed may cause the stress to decrease.

Based on Chandrashekar et al.'s (2005) study, the average ACL length for women is 26.85mm and 29.82mm for men. The graph of the results (Figure 3) shows the peak stress at ligament length of 27 mm for knee flexion of 0° and 30°, which is very close to the average ACL length for women.

3.3.2 Quadriceps Angle

The results from analyzing various quadriceps angles are shown in Table 9. The maximum principal stresses were recorded and plotted in Figure 6.

Table 9: The Recorded Maximum Principal Stresses from Analyzing Various Quadriceps Angles

Max Principal Stress (MPa)					
Quadriceps Angle (degrees)	Knee Flexion Angle (degrees)				
	0	30	60	90	120
15	0.29711	1.45650	0.00050	2.01440	1.50860
20	1.08190	1.83690	0.18289	3.34420	1.65120
25	1.15720	2.60570	1.73900	2.45320	2.04750
30	0.67422	1.41630	2.46580	2.15900	2.51040

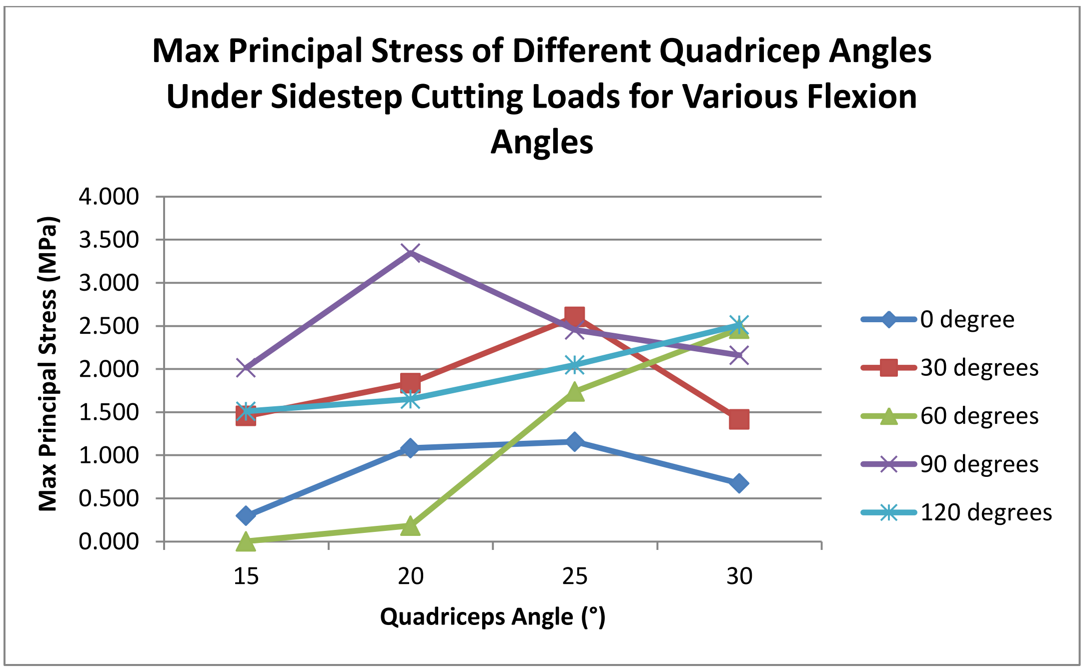

Figure 6: Graph of the Maximum Principal Stresses Recorded from Analyzing Various Quadriceps Angles at Different Flexion Angles

By observing the graphed data from the quadriceps angle test (Figure 6), noticeable increase in stress as the quadriceps angle increases can be seen. This is true for knee flexion angles at 60° and 120°. The stress at other knee flexion angles increase from quadriceps angles of 15° to 25° and decreases from 25° to 30°. The results do agree with Arendt et al. (1999) that higher quadriceps angle places the knee at greater static and dynamic stresses. However from the data collected this increase in stress is not consistent throughout all knee flexion angles. Flexion angles at 0°, 30°, and 90° experienced a decrease in stress after Q-angle of 25°. This decrease in stress at 30° Q-angle may be due to the absences of the meniscus in the knee model. The meniscus provides structural integrity to the knee especially when the knee is subjected to torsional loads. The absence of the meniscus may also alter the rest of the data by either increasing or decreasing the stress on the ACL. The data may also suggest that the 25° Q-angle is the peak of the stress experienced by the ACL and that further increasing the Q-angle lowers the stress on the ACL. Therefore based on the data acquired, the increase in stress as the Q-angle increases is present in Q-angles below 25°.

Women are known to have greater Q-angles than men. However based on how the tests are setup, the height and weight is not incorporated into the tests. The results of these tests only reflect women and men of the same height and weight. The data requires scaling according to the height and weight difference between men and women to examine the change.

4. Conclusion

ACL injuries are one of the most common knee injuries especially among athletes. The number of injuries continues to increase till this day and therefore the amount of money spent on treatment and rehabilitation continue to increase as well. Prevention methods for ACL injuries have yet been found, which is why finite element models of the ACL can be beneficial to examining the ACL without the hassle of using human cadavers.

The objective of this project was achieved as the three-dimensional finite element model was successfully constructed using the appropriate data acquired from different studies. The model was setup and able to successfully run finite element analyses with the two applied risk factors (ligament size and quadriceps angles).

Based on the results from the analyses, the following conclusions were made,

1. The data obtained evaluating this model yield stress values less than Peña et al.'s (2006) study. Difference in data may be due to the difference in models.
2. The results from analyzing ligament length showed that the ACL experienced greater stress at lower flexion angles.
3. The greatest stress was experienced at ligament length of 27mm which is the average ligament length for women.
4. The results from analyzing the quadriceps angle showed that the increase in Q-angle increases the stress on the ACL from 15° to 25° and a decrease from 25° to 30°.

4.1 Limitations

Through the course of this project there are certain limitations due to the nature of this project. These limitations include time constraint, budget constraint, and the lack of necessary equipment. Due to the nature of this project's course schedule there was a limited amount of time to complete this project. This limits the amount of testable risk factors as well as time for further improvement of the model. This project was not given a budget as it was not funded by any outside sponsors. Also the project lacked the access to necessary equipment such as medical imaging devices (MRI or CT scanners) that can help create three dimensional CAD models. This lack of equipment limited the resources that were available to work with.

4.2 Future Changes

Although the foundation for the finite element model is set, the model has yet to reach the most accurate representation of the ACL. Changes and further work is necessary to improve the model. Suggestions for future changes are as follows:

1. The model should incorporate all necessary anatomical components of the knee. These structures include the two other knee ligaments (MCL & LCL), the meniscus, patella, and patellar tendon as these structures may contribute to the stress distributions within the knee.
2. The model should incorporate muscle loads of the lower extremity that contribute to the movement of the knee joint. The activation of the muscle may impact the stresses on the ACL.
3. The model should replace the ligaments with ones obtained from MRI or CT scan. This will improve the accuracy of the ligament geometry and yield more accurate results.
4. The model should be scaled according to gender by incorporating the height and weight.

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Appendix A

Study	Description
Limbert et al. 2004	A three dimensional finite element model of the ACL was developed and simulations of passive knee flexion were performed. The goal of this study was to assess the performance of a 3D finite element model of the ACL, by comparing the predicted resultant force generated by the ligament with those reported in the literature. The result maximum principal stresses were recorded (Figure 9) at different knee flexion angles. Their developed model was able to reproduce the qualitative mechanical behaviors of the ACL and the quantitative data from cadaver testing.
Park et al. 2010	The goal of this study was to develop a FEM of the ACL to conduct finite element analysis (FEA) on the ACL impingement against the intercondylar notch under tibial external rotation and abduction. The FEA showed that impingement between the ligament and the lateral wall of intercondylar notch could occur when the knee at 45° was externally rotated at 29.1° and abducted at 10.0°. Their results showed that the impingement force increased as the abduction and external rotation increased.
Peña et al. 2006	This study developed a three dimensional FEM of the human knee that includes the femur, tibia, articular cartilage, menisci and ligaments (patellar tendon, anterior cruciate, posterior cruciate, medial collateral and lateral collateral). Their model was used to study the kinematics and stresses. A combination of 1150 N compressive load, 10 Nm valgus torque and a 134 N anterior load were applied to their model. The FEM and the maximum principal stresses in the ligaments are shown in Figure 11.
Song et al. 2004	Studied the force and stress distribution on the ACL with a 134N anterior tibial load at full extension, and their goal was to determine the feasibility of developing a FEM of the human ACL. This goal was accomplished as the FEM was developed, validated and the force and stress distribution within the ACL was determined. They found that the stress distribution on the ACL during a 134N anterior tibial load at full extension was not uniform throughout the ACL. The highest stress was found to be located near the femoral intersection area as the ACL wraps around the bone when load is applied.

Table 10: Studies Done Using Finite Element Models

Appendix B

Risk Factors	Risk Factors That Will Be Tested in This Model	Studies Done Relating to Risk Factor and ACL Injuries	Conclusion from Studies
Quadriceps Angle (“Q” angle)	X	Alentorn-Geli et al. 2009	Aletorn-Geli et al. studied the prevention of noncontact ACL injuries in soccer players. Their study reviewed different risk factors of ACL injuries. In Aletorn-Geli et al.’s review on quadriceps angle, based on the current research that have been done pertaining the relationship between quadriceps angle and ACL injuries, there isn’t enough evidence to suggest that increase in Q-angle increases ACL injuries.
Intercondylar notch size		Souryal 1993	Souryal studied the relation between intercondylar notch size and ACL injuries in athletes. Intercondylar notch size and ACL injury information from 902 high school athletes were recorded and examined. Souryal concluded that athletes with a stenotic intercondylar notch are at significantly greater risk of sustaining noncontact ACL injury.
Knee Joint Laxity		Myer et al. 2008	Myer et al. studied the effects of joint laxity on the risk of ACL injury in female athletes. The purpose of their study is to determine if the increase in knee joint laxity would increase the risk of ACL injury. Their results indicate that increasing joint laxity may contribute to increased risk in ACL injury.
Ligament size	X	Anderson et al. 2001	Anderson et al. studied the correlation of ligament size and muscle strength to gender differences in ACL injuries. 50 male and 50 female athletes were evaluated and from their results, it showed that female ACLs are smaller than male ACLs even with adjustments for body weight. Therefore the difference in ligament size is exposing females to greater ACL injury rates.
Hormone Levels		Wojtys et al. 2002	Wojtys et al. studied and recorded 69 female athletes and their menstrual cycle details, hormone levels, and recorded the menstrual cycle phase in which ACL injuries occur. The results from this study indicated that women had significantly greater than expected percentage of ACL injuries during mid-cycle (ovulatory phase) and a less than expected percentage of injuries during the luteal phase. These results suggest that women are more

			susceptible to ACL injuries during the ovulatory phase of the menstrual cycle.
Muscle Strength		Anderson et al. 2001	The results from their muscle strength study showed that male athletes statistically had greater quadriceps and hamstring muscle strength than female athletes. Also their study showed that stiffness and muscular strength increase stress on the ACL in female athletes.
Level of Conditioning		Hewett et al. 1999	Hewett et al. studied the effect of neuromuscular training on knee injuries in female athletes. Two groups of female athletes were monitored. One group was trained and the other group did not undergo training. The results of their study showed a decrease in incidences of knee injuries in female athletes.
Motor Control Strategies		Zebis et al. 2008	Zebis et al. studied the effects of neuromuscular training on the knee joint motor control during side-cutting maneuver in female athletes. The study aimed to analyze neuromuscular adaptation mechanisms during side-cutting maneuver which is known to be associated with noncontact ACL injuries. The results from their study showed that neuromuscular training increased activity in the hamstring muscles, therefore decreasing the risk of dynamic valgus. The neuromuscular adaptation during a side-cutting maneuver can potentially reduce the risk of noncontact ACL injuries.
Skill Level		Harmon & Dick 1998	Harmon and Dick studied the relationship between skill level and ACL injuries. Skill level and injury information was recorded from women and men's basketball and soccer players of different divisions. Their results showed no relationship of ACL injury rate to division level in both men and women's basketball or soccer. Therefore, skill level was not found to relate to ACL injury rate in men or women.
Shoe-Surface Interface		Alentorn-Geli et al. 2009	From the review on shoe-surface interface, based on the type of surface athletes play on have an impact on ACL incidence rates among athletes. They mention that artificial surface is associated with higher shoe-surface traction, therefore greater risks for ACL injuries. Factors that influence shoe-surface traction consist of ground hardness, ground coefficient of friction, ground dryness, grass cover, root density, length of cleats on shoes, and game pace.

Table 11: List of Risk Factors and Risk Factor Related Studies

Appendix C

Model Troubleshooting

Finite element analysis was not successful using the femur, tibia, and patella bone models provided by the National Library of Medicine. The FEA software used (ANSYS) failed to mesh the assembled model resulting in unsuccessful analysis using this assembled model. Without a meshed model, analysis cannot be done. Multiple attempts were made to solve this issue however this issue was not solved. The following explains the setup and settings used for the model and the description of attempts made. The explanation of this issue may help future researchers in avoiding this type of issue and can help in identifying a solution to this problem.

Model used: Femur, tibia and patella from the National Library of Medicine
Model file formats used: STEP, IGES, ProENGINEER, SolidWorks
Finite element software used: ANSYS Workbench version 13
Settings Set:

- Mesh Settings:
  - Relevance Center = Coarse
  - Element Size = Default
  - Initial Size Seed = Active Assembly
  - Smoothing = Low
  - Transition = Fast
  - Span Angle Center = Coarse
- Mesh Option Setting:
  - Mesh Method = Automatic

Steps taken to analyze model:

1. Input assembled CAD model into ANSYS Workbench
2. Set material properties
   a. The material properties were assigned and set according to the values reported from Ozkaya & Nordin (1999), shown in Table 3.
3. Set mechanical properties
   a. The mechanical properties were assigned and set according to the values reported from Chandrashekar et al. (2006), shown in Table 4.
4. Set contact zones
   a. Three contact zones were set. First is a frictionless contact zone set between the base of the femur with the cranial portion of the tibia. Second is a bonded contact zone between the ACL and the femoral insertion site. The third is also a bonded contact zone between the ACL and the tibial insertion site.
5. Set boundary conditions
   a. The femur was set as a fixed structure while the tibia is free to move in the flexion/extension plane, varus/valgus, and internal/external rotations.
6. Add forces
   a. A 134N anterior tibial load was applied to the tibia.
7. Set solutions to be returned
   a. The analysis was set to solve for maximum stress.
8. Attempt to solve

a. As the software attempts to generate a solution, a mesh is first generated for the model. The analysis stops as it is unable to generate a mesh, the following error messages are returned:

"A mesh could not be generated using the current meshing options and settings"
"The following surfaces cannot be meshed with acceptable quality. Try using different element size or virtual topology."

Attempts made to fix problem:

1. Changing element size
    a. Element size was varied from small element sizes to bigger element sizes
2. Changing meshing options
    a. Meshing options were changed to tetrahedrals
3. Changing file format
    a. Various formats were used, such as igs, iges, SolidWorks, ProENGINEER
4. Changing combinations of bone models in assembly
    a. Various combinations of assemblies were used to narrow down the model that may be causing the error. Combinations such as femur & tibia, femur & ACL, tibia & ACL all did not work. ANSYS was only able to mesh the modeled ACL but not the tibia, femur or patella.

Observations made from attempts:

1. When importing the assembly into ANSYS, partial surfaces are missing from the bone models when using ProENGINEER, SolidWorks, and igs formats.
2. When individual bone models are imported into ANSYS for analysis, no error message occurs, meshing and analysis were successful. Only unsuccessful when more than one model is imported.

Hypothesis on this issue:

There may be a problem with the bone model file itself. There may be in gaps within the models that are preventing the software to mesh the geometry properly.

Appendix D

Within McLean et al.'s (2003) study in developing a 3D model to predict knee joint loading, they were able to determine the muscle groups that are involved in a sidestep cutting maneuver. These muscle groups and their corresponding maximum isometric force are shown in Table 12.

Table 12: Muscle Group Activation During Sidestep Cutting Maneuver (McLean et al. 2003)

Muscle (group)	F_{max} (N)
Rectus femoris	780
Vastus lateralis	1870
Vastus medialis	1295
Vastus interm	1235
Biceps femoris LH	720
Biceps femoris SH	400
Hamstrings	1360
Gastrocnemius	1605
Soleus	2830
Tibialis posterior	1270
Tibialis anterior	600